Sensors and Actuators B 207 (2015) 1010–1018

Contents lists available at ScienceDirect

Sensors and Actuators B: Chemical

journal homepage: www.elsevier.com/locate/snb

Effect of surface conductivity on the sensitivity of interdigitated impedimetric sensors and their design considerations

A. Guimerà[a,b,**], G. Gabriel[a,b], E. Prats-Alfonso[a,b], N. Abramova[c], A. Bratov[a,*], R. Villa[a,b]

[a] Institut de Microelectrònica de Barcelona, IMB-CNM (CSIC), Campus UAB, 08193 Barcelona, Spain
[b] Centro de Investigacion Biomédica en Red, Biomateriales y Nanomedicina (CIBER-BBN), Spain
[c] ITMO University, Kronverkskiy pr, 49, 197101, St. Petersburg, Russia

ARTICLE INFO

Article history:
Available online 7 November 2014

Keywords:
Interdigitated electrodes
Impedimetric biosensor
FEM modeling
Surface conductivity
Electrical double layer

ABSTRACT

The use of impedimetric biosensors based on interdigitated electrodes (IDEs) for biochemical applications has gained increased interest during the last two decades. Recently a concept of a 3D-IDEs with insulating barriers separating electrodes digits was introduced. This geometry enhances the sensitivity of the transducer to biochemical reactions at the sensor surface. Although there is an experimental evidence of this increase in the sensitivity, its quantification is currently unstudied. In this work the impact of the geometry on sensitivity of IDEs impedimetric biosensors based on surface conductivity variations is studied. For that a finite elements method (FEM) model that takes into account the surface conductivity phenomena has been developed. Its feasibility has been validated comparing modeled and experimental results obtained in immunochemical reaction of IgG protein. Using this FEM model the increase of sensitivity produced by the presence of a dielectric barrier between the electrodes has been quantified. Moreover, the biosensor response has also been analyzed taking into account different geometrical parameters such as electrode width or separation. Finally, this work proposes the design guidelines for maximizing and improving the sensitivity of IDEs based biosensors.

1. Introduction

Electrochemical biosensors based on impedimetric transduction are becoming of great interest due to their ability to perform label-free detection [1,2]. These devices provide viable solutions for the monitoring of a wide range of analytes with several advantages over other techniques, such as small size, low cost, fast response and analyte determination without any additional markers. Moreover, in contrast with other electrochemical transducers, like amperometric or potentiometric, no reference electrode is required which simplifies the measurement and permits sensor miniaturization. In the recent years a great effort has been made in this research area to understand the fundamentals of these biosensors and achieve really usable analytical devices with a higher sensitivity and accuracy, better reproducibility, lower detection limit [3]. A wide range of different impedimetric biosensors was reported and the number of publications grows from year to year. Impedimetric biosensors may be used to register changes in the electrical properties at the electrode surface modified with bio-receptor molecules. Biochemical reactions at the sensor surface may be registered as capacitance changes in non-Faradaic measurements [4–6]. In the presence of redox active species in test solution Faradaic charge transfer resistance [7,8] is measured, which may be affected by interactions of a target biomolecule with a probe-functionalized sensor surface.

In the case of interdigitated electrode arrays planar microband electrodes between which the impedance is measured are very closely situated, so that in this case changes in electrical properties of the interdigital space may affect the sensor impedance.

The use of interdigitated electrodes (IDE) for biochemical sensing applications has gained increased interest during the last two decades making IDEs one of the most commonly used electrochemical sensor structures [8–11]. Recently, Bratov et al. introduced the concept of a three-dimensional (3D)-IDE with electrode fingers separated by an insulating barrier made of silicon dioxide (SiO_2) [12,13]. The specific design of this sensor structure permits to enhance its sensitivity for biochemical reactions taking place at the sensor surface [14]. On the contrary, Rana et al. [15] presented a comparison between planar and 3D-IDE based on an analytical method and finite element analysis. It was concluded that the introduction of a dielectric barrier does not increase the sensitivity of a 3D-IDE for impedance changes caused by the additional

* Corresponding author. Tel.: +34 93 594 77 00; fax: +34 935 80 14 96.
** Corresponding author at: Institut de Microelectrònica de Barcelona, IMB-CNM (CSIC), Campus UAB, 08193 Barcelona, Spain.
E-mail addresses: anton.guimera@csic.es (A. Guimerà), andrei.bratov@imb-cnm.csic.es (A. Bratov).

http://dx.doi.org/10.1016/j.snb.2014.10.134
0925-4005/

insulating layer (imitating a molecular layer) but only offers more surface area for molecular immobilization. However, used model did not take into account the presence of a surface conductive layer that is responsible for higher sensitivity of 3D-IDE devices [16,17].

Different strategies may be applied for constructing an affinity biosensor based on an IDE transducer. Their functionalization with probe molecules (e.g. antibody, single strand DNA) may be done only on the metallic electrodes, e.g. gold, of the IDE [18–20]. However, in this case advantages offered by a specific IDE geometry are not used as, once again, a surface capacitance or a charge transfer resistance is determined. On the contrary, when all the sensor surface comprising electrodes and the interdigital space or only interdigital space is covered by bio-receptor molecules [21,22] changes in electrical charge distribution due to the surface biochemical reactions may significantly affect the impedance in the vicinity of the IDE sensor surface. Therefore, depending on the geometry and the functionalization method we can distinguish different configuration designs that are presented in Fig. 1 for planar IDEs (A and B) and for 3D-IDEs (C and D).

In the literature one finds a large variety of experimental strategies related to application of planar or 3D-IDEs along with the lack of theoretical background regarding the processes governing the response of these sensors and main motivation of the present work was to bridge this gap.

Here a finite elements method (FEM) is used to model the current distribution and response of planar and 3D IDEs sensors in order to demonstrate and quantify experimentally observed changes in their sensitivity. Essential difference in our simulation strategy in comparison with the work [15] is the presence of a surface conductivity layer that affects the IDEs behavior especially when working in low conductivity solutions. The work is focused on the study of the IDEs behavior in response to non-Faradaic reactions on its surface. In the first part of the work the FEM model is experimentally validated for a 3D IDEs transducer registering antigen–antibody interactions of the Immunoglobulin G (IgG) antigen-antibody pair, one of the most standard reactions in immunochemistry. In the second part of the work this model is implemented to study how the sensitivity of IDEs depends on the sensor configuration (Fig. 1 Conf. A–D) with different combinations of geometric parameters (gap between electrodes (Gap), electrode width (W_e) and height of the dielectric barrier ($H_{barrier}$)). Finally, conclusions regarding the sensitivity of each of the commented configurations are presented.

2. Materials and methods

2.1. Experimental model

2.1.1. 3D-IDEs design and fabrication

The device was fabricated using conventional microelectronic techniques as presented earlier [12,13]. Silicon wafer covered with a thermally grown silicon dioxide layer of 2500 nm was used as a substrate. A 230 nm thick layer of tantalum silicide ($TaSi_2$), which is a highly conductive material, is deposited by magnetron sputtering. The first photolithographic step defines collector bars and digits of two electrodes. The patterning is done by a reactive ion etching technique. This results in an interdigitated electrode array with 216 digits of 3 μm width and 3 μm gap between the adjacent electrode digits. The electrode digits are 1.5 mm long and the aperture between the adjacent digits is 1.4 mm.

To form the contact pads 1 μm of aluminum is deposited and patterned using standard photolithographic and etching steps leaving metal only at extremes of the two collector bars. In the final step the wafer with formed IDE devices is covered with a 4 μm thick silicon oxide layer deposited by a low pressure chemical vapor deposition (LPCVD). This material is virtually the SiO_2 but with a lower density compared to thermally grown silicon oxide or quartz. Photolithography is used to define the trenches to be opened in the oxide layer over the electrodes digits and over contact pads. These zones are opened by deep reactive ion etching (DRIE), which permits to obtain nearly vertical walls. In this way capillaries are formed in silicon dioxide over the electrode digits. The capillaries over each electrode digit are 3 μm wide, 4 μm high, 1.5 mm long and are opened at their top.

After being cut from the wafer the sensors are glued to a printed circuit board (PCB) substrate with copper leads and are wire bonded for electrical connections. Contact pads and wires were encapsulated using epoxy resin.

2.1.2. Chemicals and reagents

3-Glycidoxypropyltrimethoxysilane, carbonate–bicarbonate buffer solution capsules, monoclonal anti-goat/sheep IgG (clone GT-34 produced in mouse and IgG protein from goat serum) were purchased from Sigma-Aldrich (Milwaukee, WI, USA); phosphate buffered solution tablets was obtained from GIBCO Invitrogen (Barcelona, Spain), standard conductivity solution with 1278 μS/cm at 20 °C (KCl 0.01 M), ethanol 99.5% and sodium hydroxide were procured from Panreac (Barcelona, Spain), and hydrogen peroxide 30% (v/v) was obtained from Fisher Scientific (Waltham, MA, USA).

2.1.3. Immobilization of antibody and protein detection

IDE sensors were cleaned with ethanol and deionized (DI) water and were immersed in a NaOH 1 M for 30 min in order to enrich the surface with silanol groups. After the activation step the chips were rinsed with DI water and ethanol and dried again with nitrogen. The chips were exposed to 3-glycidoxypropyltrimethoxysilane 2.5% volumetric ethanol solution for 16 h at room temperature.

The terminal epoxy group of silane attached to the surface of the SiO_2 readily reacts with amino groups of biomolecules thus fixing them on the surface. In this way immobilization of the IgG antibody (1.0 μg mL^{-1} in carbonate buffer, pH 9.6) overnight at 4 °C was performed. The sensors were rinsed with PBS buffer and dried with a N_2 stream. Reaction with IgG protein for the final detection was performed by immersing the chips into a 5 μg mL^{-1} protein IgG solution in PBS for 30 min at room temperature. In the control experiment the sensor modified with the antibody was kept in the PBS solution that did not contain IgG protein. These sensors are referred further in the text as Native – SiO_2 surface without any modification, AbIgG – modified with antibodies, protIgG – after reaction with protein, and control – not exposed to protein.

X-ray photoelectron spectroscopy (XPS) experiments showed that tantalum silicide surface is covered with a thin 0.2–0.5 nm layer of a native oxide formed at room temperature. Therefore, chemical modification with silane, and subsequent antibody immobilization, goes not only on the SiO_2 barrier surface but also on the surface of the electrode digits (the case presented in Fig. 1D).

2.1.4. Electrochemical impedance spectroscopy

Electrochemical impedance spectroscopy (EIS) was used to characterize fabricated sensors, as well as to measure the response of immunosensors. EIS measurements were carried out at open circuit potential without external biasing in the frequency range 10–10^5 Hz with 25 mV amplitude of the test signal using an impedance analyzer Solartron 1260A and a Solartron® 1287A potentiostat used as a front-end to increase the input impedance. Conductivity of the solutions was controlled using a commercial Crison micro CM2202 conductimeter.

EIS characterization of sensors with native surface, after antibody immobilization and after reaction with protein was performed in KCl water solutions with six different conductivities.